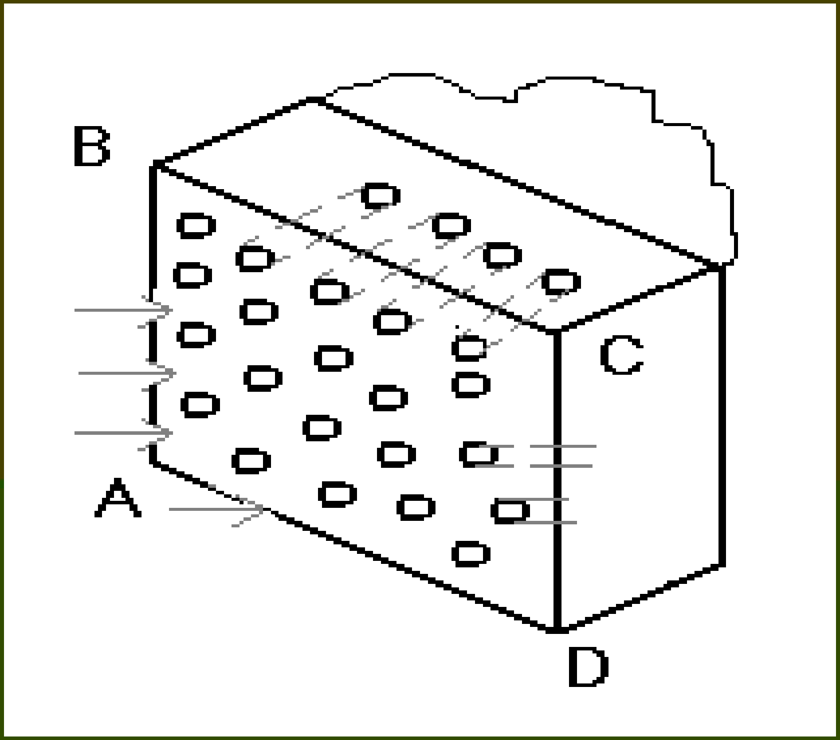
PRODUCTION OF RADIOISOTOPES

Activation of Isotopes

When almost any material is placed in a reactor, it may become activated by neutron bombardment. The probability of activation is determined by the cross section for the nuclear reaction. It is usually represented by σ and is expressed in cm^2 per atom or in barns per atom where 1 barn=10^{-24} cm^2.

If a neutron passes through this area, that is, “hits” the nucleus, then it is captured and an activation takes place.

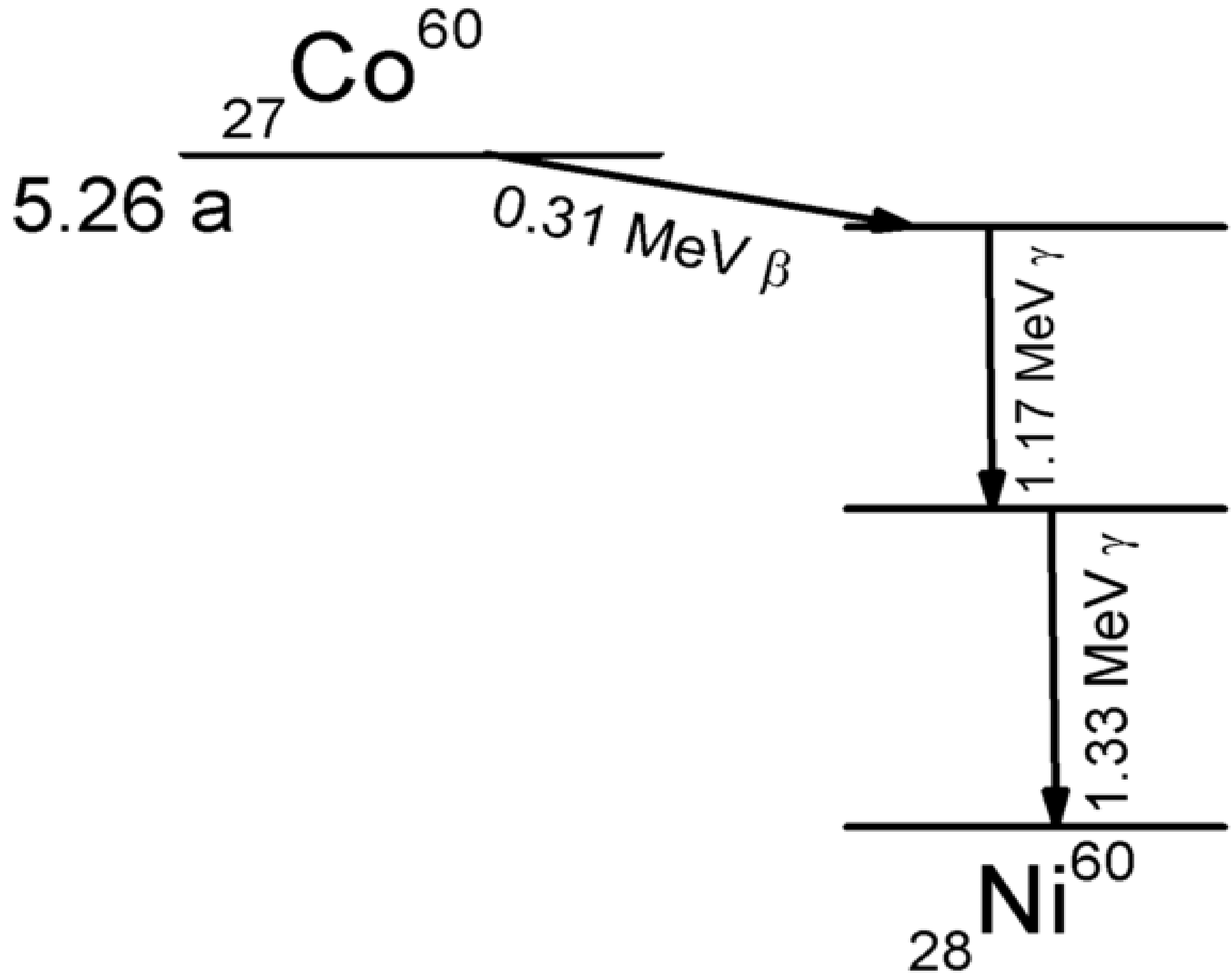
If a neutron flux φ ($n.cm^{-2}.s^{-1}$) is incident on a target material contains N_t as shown in figure



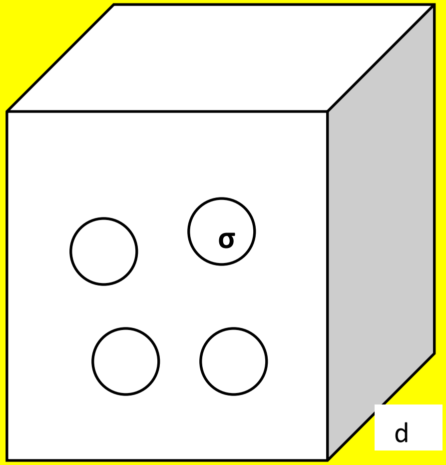
Consider all the parts of the target is exposed to the neutron flux. To calculate the number of nuclides that react with neutrons to produce radio nuclides ,assume that the cross sectional area of the taget nucleus is

$$\sigma$$

1 barn = 10 $^{-24}$ cm^2



$^{59}Co + n \text{ ---> } ^{60}Co + \gamma$ (Q = 7.492 MeV)



$AQ\sigma/A$ = probability for one falling neutron to make a reaction. Q no of nuclei per unit area When (n) neutrons fall on the target , the no of reactions = n Q σ

$$N = (\Phi A \Delta \boldsymbol{t})\boldsymbol{\sigma Q} = (\Phi A \Delta \boldsymbol{t})\boldsymbol{\sigma N_V d} = (\Phi \Delta \boldsymbol{t})\boldsymbol{\sigma N_V A d}$$
$$= (\Phi \Delta \boldsymbol{t})\boldsymbol{\sigma N_t}$$

Example

Find the number of active atoms of CO-60 produced in a 1g sample of CO-59 that is placed in a neutron flux density of $10^{13} cm^{-2} s^{-1}$ for 1y ($3.16 * 10^7$ s). The atomic weight of cobalt is 58.94, and the activation cross section is 37 barns per atom.

$$\text{Number atoms in target} = \frac{6.07 * 10^{23}}{58.94} g^{-1} * 1g = 1.02 * 10^{22}$$

No. activated = $1.02*10^{22}*37*10^{-24}*10^{13}*3.16*10^{7}$ = $1.19*10^{20}$

In deriving the equation the implicit assumption has been made that nuclei at the front of the target do not shield those at the back and this is often the case.

Another assumption we have made is that in the activation process the number of target atoms remains constant during the irradiation. In example this is essentially true since only a little over 1 percent of the atoms present initially are activated in the one year’s irradiation.

Activity Produced by Neutron Irradiation

In the activation of isotopes, one is usually not interested in the number of active atoms produced but in the activity of the sample that result from an irradiation. Let Δλ represent the activity resulting from the production of ΔN activated nuclei according to equation. If λ is the decay constant of the isotope that is produced then

$$\Delta A = \lambda \Delta n = \lambda N_t \sigma \varphi \Delta t = \frac{0.693}{T} N_t \sigma \varphi \Delta t$$

To illustrate this we calculate the activity produced in the previous example :

$$\Delta A = \frac{0.693}{5.26y} * 1.19 * 10^{20} = 1.57 * 10^{19} y^{-1}$$

$$= \frac{1.57 * 10^{19}}{3.16 * 10^{7}} = 4.96 * 10^{11} Bq = 13.4ci$$

In this calculation we have taken no account of the decay of the isotope during the irradiation. To take this into account we may proceed as follow. At any time t during the irradiation, let the number of activated atoms be N.

In the interval of time, Δt, a number $N_1 \sigma \varphi \Delta t$ will be produced and a number $\lambda N \Delta t$ will decay, hence the increase in number of active atom ΔN will be:

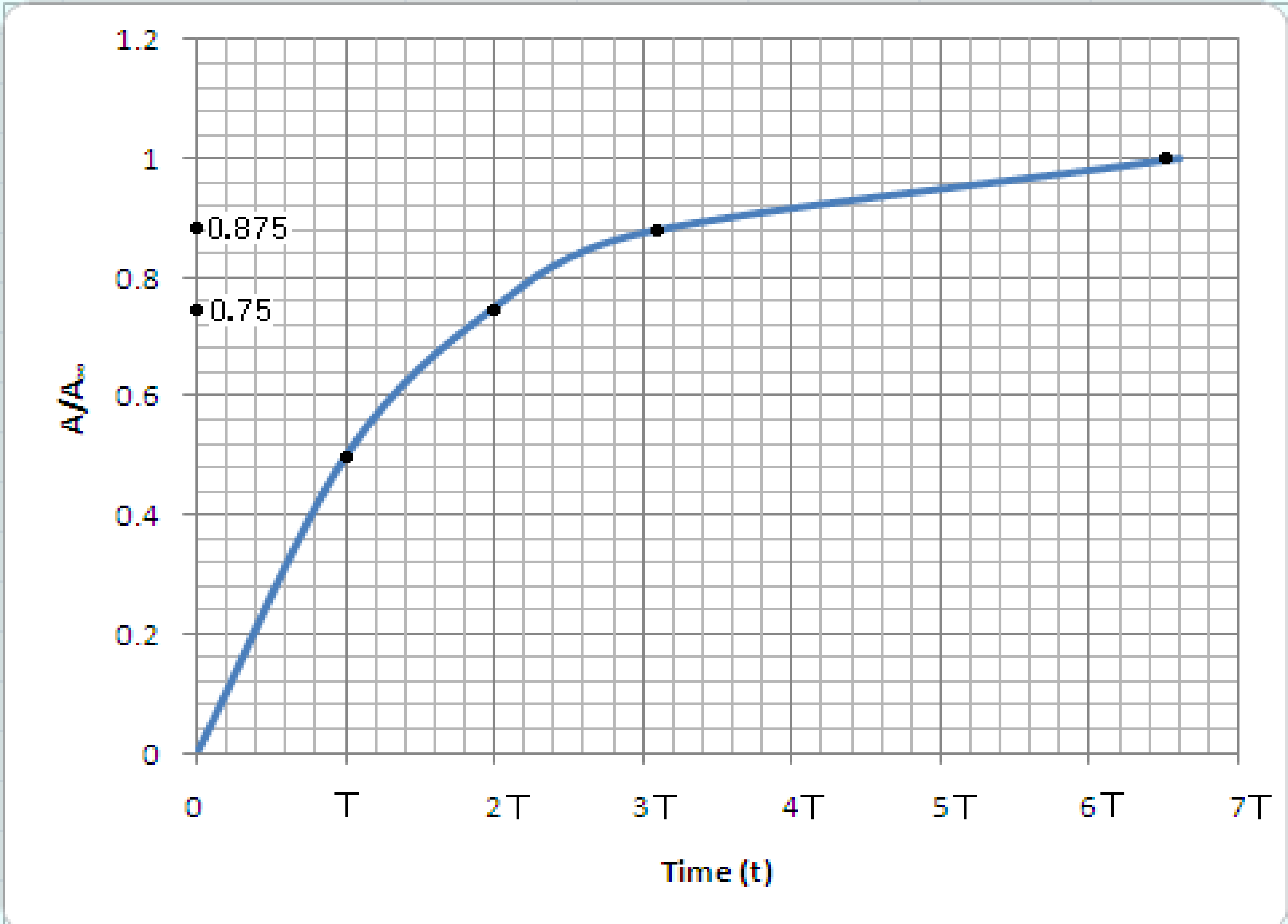
$$\Delta N = \lambda N_t \sigma \varphi \Delta t - \lambda N \Delta t$$

$$\frac{dN}{dt} = \lambda N_t \varphi \sigma - \lambda N$$

$$N(t) = \frac{N_t \sigma \Phi}{\lambda}\left(1 - e^{-\lambda t}\right)$$

$$\lambda N = N_t \sigma \Phi \left(1 - e^{-\lambda t}\right)$$

$$A = A_\infty \left(1 - e^{-\lambda t}\right)$$



1.2
1
0.8
0.6
0.4
0.2
0
0.875
0.75
A/A∞
0
T
2T
3T
4T
5T
6T
7T
Time (t)

Problem

Find the saturation activity in the previous example.

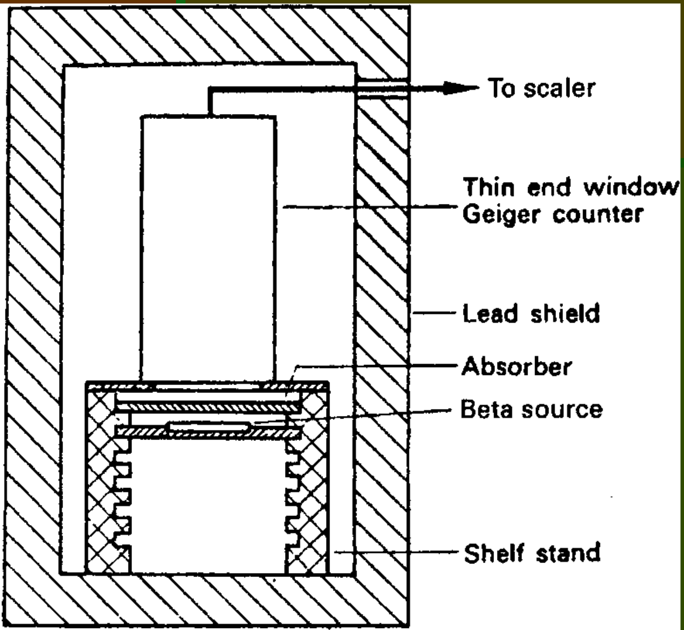
What is the activity of the produced radioisotope after activation time equal the half life time of the produced Co-60.

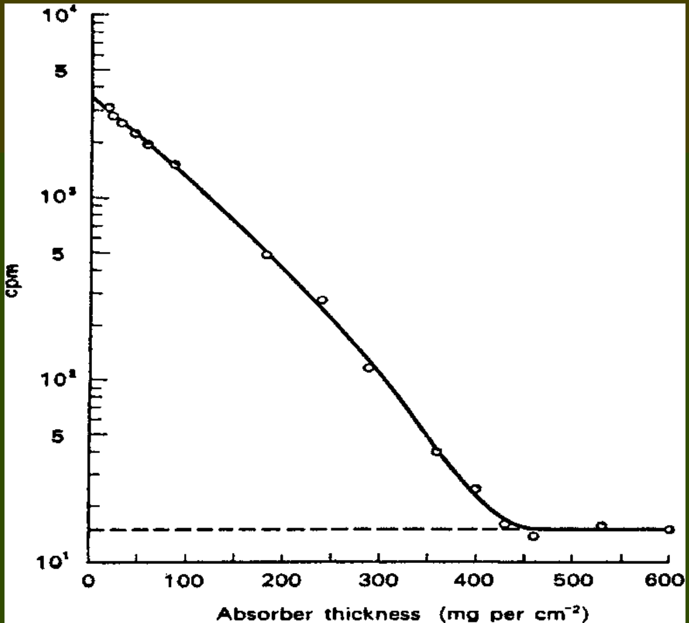
INTERACTION OF RADIATION WITH MATTER

Beta Rays

The attenuation of beta rays by any given absorber may be measured by interposing successively thicker absorbers between a beta ray source and a suitable beta-ray detector, such as a Geiger counter

When this is done with a pure beta emitter, it is found that the beta- particle counting rate decreases rapidly at first, and then slowly as the absorber thickness increases. Eventually, a thickness of absorber is reached that stops all the beta particles





Detailed analysis of experimental data shows that the ability to absorb energy from beta rays depends mainly on the number of absorbing electrons beta ray, that is. On the areal density (electrons per cm^2) of electrons in the absorber, and, to a very much lesser degree, on the atomic number of the absorber.

Interaction between the electric fields of a beta particle and the orbital electrons of the absorbing medium leads to electric excitation and ionization. The electron is held in the atom by electrical forces, and energy is lost by the beta particle in overcoming these forces. Since electrical forces act over long distance, the “collision” between a beta particle and an electron occurs without the two particles coming into actual contact as in the case of the collision between like poles two magnets.

The amount of energy lost by the beta particle depends on its distance of approach to the electron and on its kinetic energy. If ϵ is the ionization potential of the absorbing medium and E_t is the energy lost by the beta particle during the collision, the kinetic energy of the ejected electron, E_k, is:

$$E_k = E_t - \epsilon$$

In many ionization collisions, only one ion pair is produced. In other cases, the ejected electron may have sufficient kinetic energy to produce a small cluster of several ionizations; and in a small proportion of the collision the ejected electron may receive a considerable amount of energy, enough to cause it to travel a long distance and to leave a trail of ionizations.

Such an electron, whose kinetic energy may be one the order of 100eV, is called a delta ray.

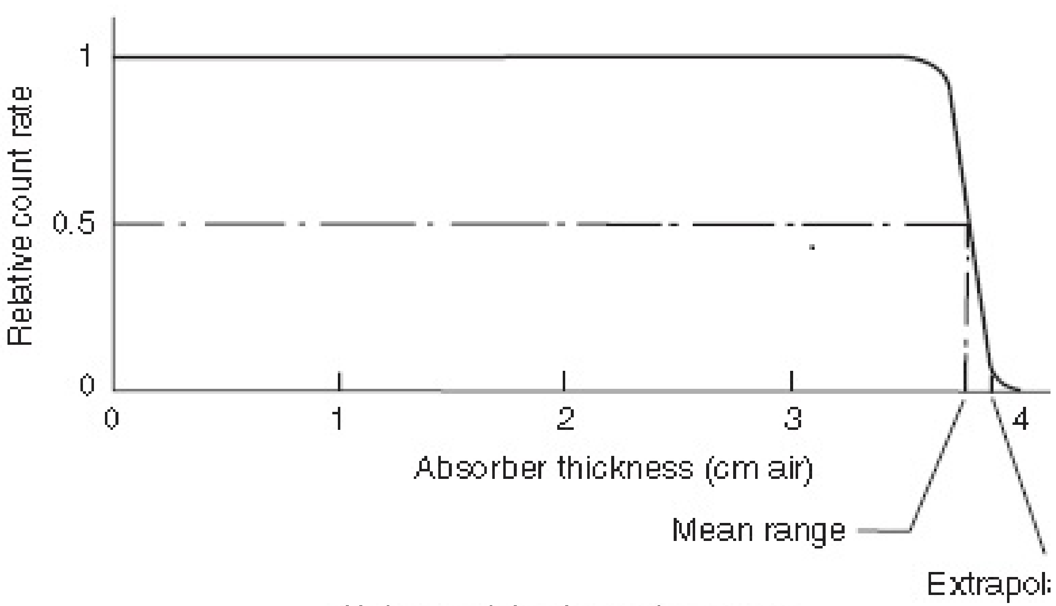
Alpha Rays

Alpha rays are the least penetrating of the radiations. In air, even the most energetic alphas from radioactive substances travel only several centimeters, while in tissue, the range of alpha radiation is measured in micron ($1\mu = 10^{-4}$cm). The term range, in the case of alpha particles, may have two different definitions: mean range and extrapolated range.

The difference between these two ranges can be seen in alpha particle absorption curve. An alpha particle absorption curve is flat because alpha radiation is essentially monoenergetic. Increasing thickness of absorption serves merely to reduce the energy of the alphas that pass through the absorbers; the number of alpha is not reduced until the approximate range is reached.

At this point, there is a sharp decrease in the number of alphas that pass through the absorber. Near the very end of the curve, absorption rate decreases due to straggling, or the combined effects of the statistical distribution of the " average" energy loss per ion and the scattering by the absorber nuclei.

The mean range is the range most accurately determined, and corresponds to the range of the "average" alpha particle. The extrapolated range is obtained by extrapolating the absorption curve to zero alpha particles transmitted.



Alpha-particle absorption curve.

Photoelectric absorption

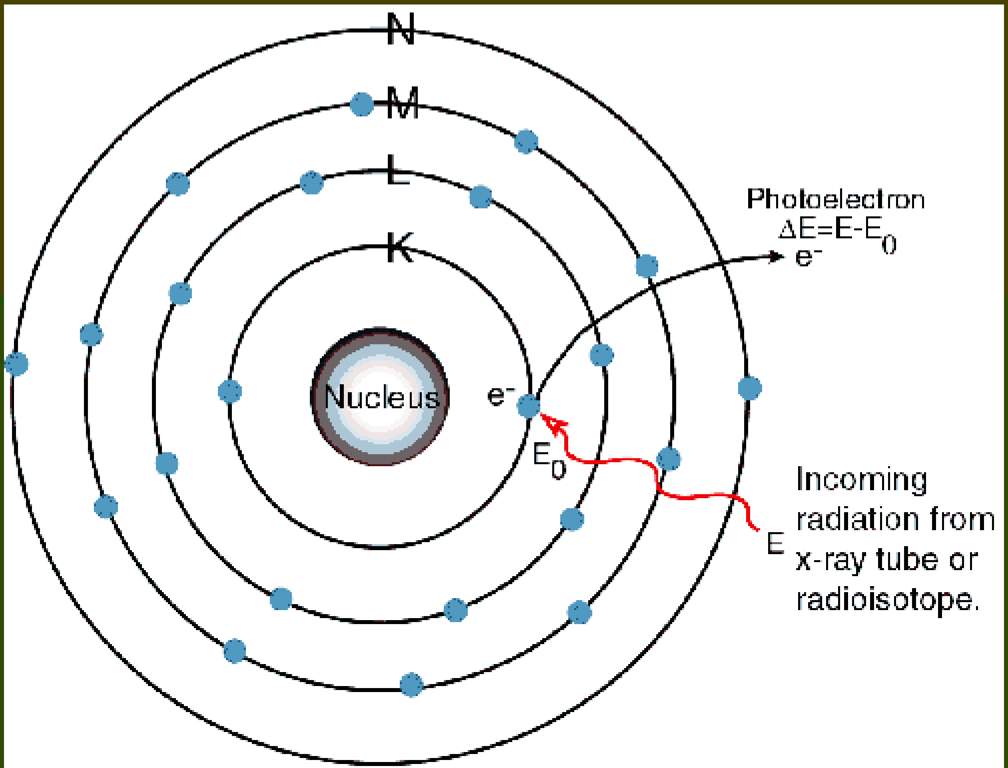
The photoelectric effect, in which the photon disappears, is an interaction between a the primary ionizing particle resulting from this interaction is the photoelectron,
whose energy is given by equation

$E_{pe} = hf - w$

E_{pe}=kinetic energy of the emitted electron

f = frequency of the incident photon

w=binding energy of the electron (work function)

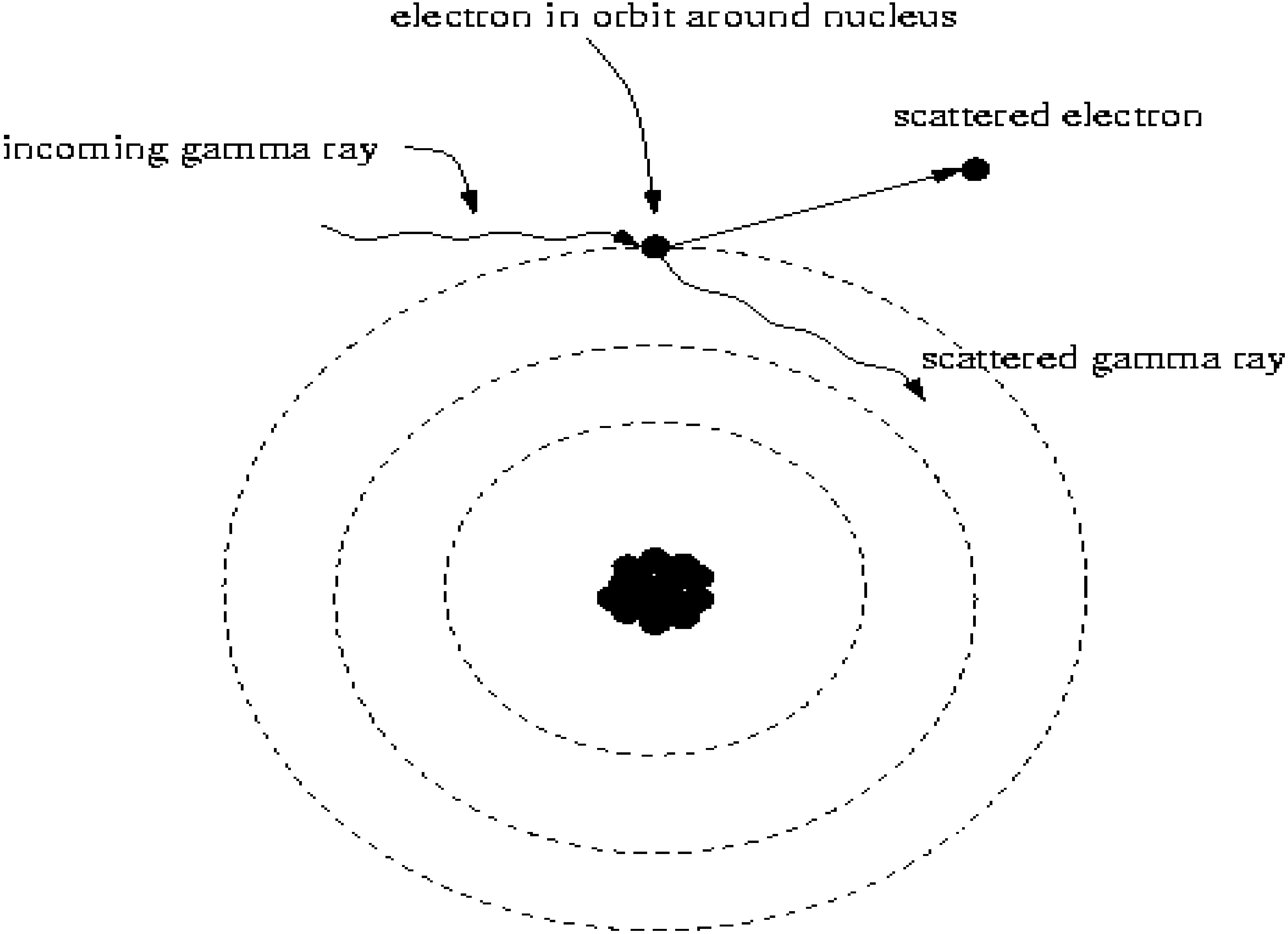


The X-ray photon dissipates its energy through collision with an atom in the absorbing medium . An inner electron encounters the photon and absorbing its energy, the photon disappears with the ejection of the electron. In this interaction the energy of the incident photon must be higher than the binding energy of the electron(*w*) . The excess energy is used by the emitted electron as kinetic energy E_{pe}. The <u>*photoelectric effect*</u> is favored by low energy photons and high atomic numbered absorbers.

Compton Effect :

This mechanism is predominant for photons with medium energies (E> 0.5 Mev) where the incident photons with energy hf_0 interact with the outer most electrons (free electrons)of the target atoms whose binding is nearly zero. During the collision part of the energy of the photon is transferred to the free electron to be used by the electron as kinetic energy. E_{CS} , the photon with reduced energy is scattered with less energy hf' . The ejected electron is known as a recoil electron .

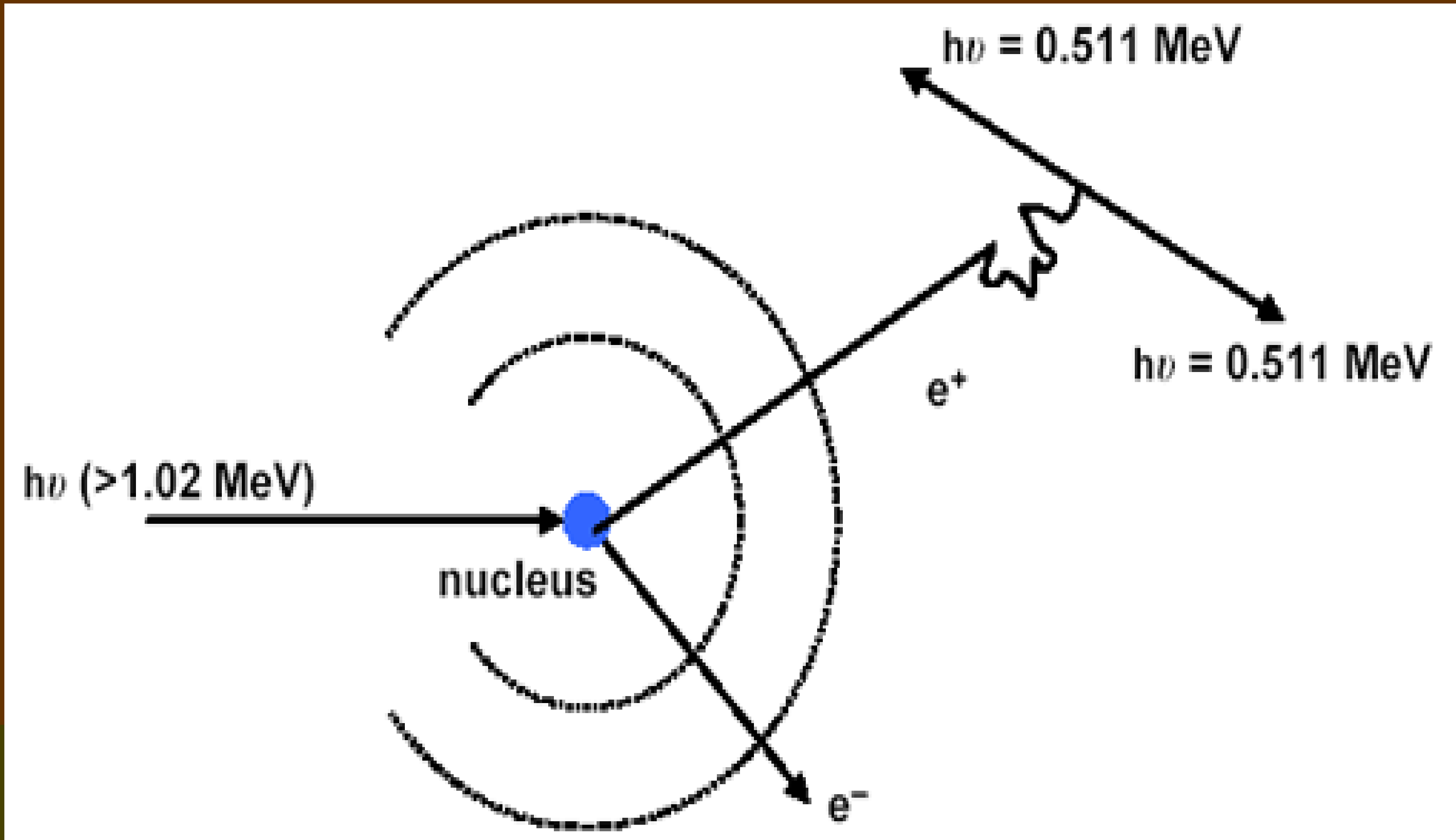
$$E_{CS} = hf_o - hf'$$



Compton Scattering:

Pair Production:

The photon can interact with the nucleus of an atom. The photon disappears and an electron and positron emerge. It is required that the energy of the incident photon is to be higher than *1.02Mev* . *1.02Mev* will be used to creat the two electrons and the energy difference will be equally distributed between the two electrons as kinetic energy.

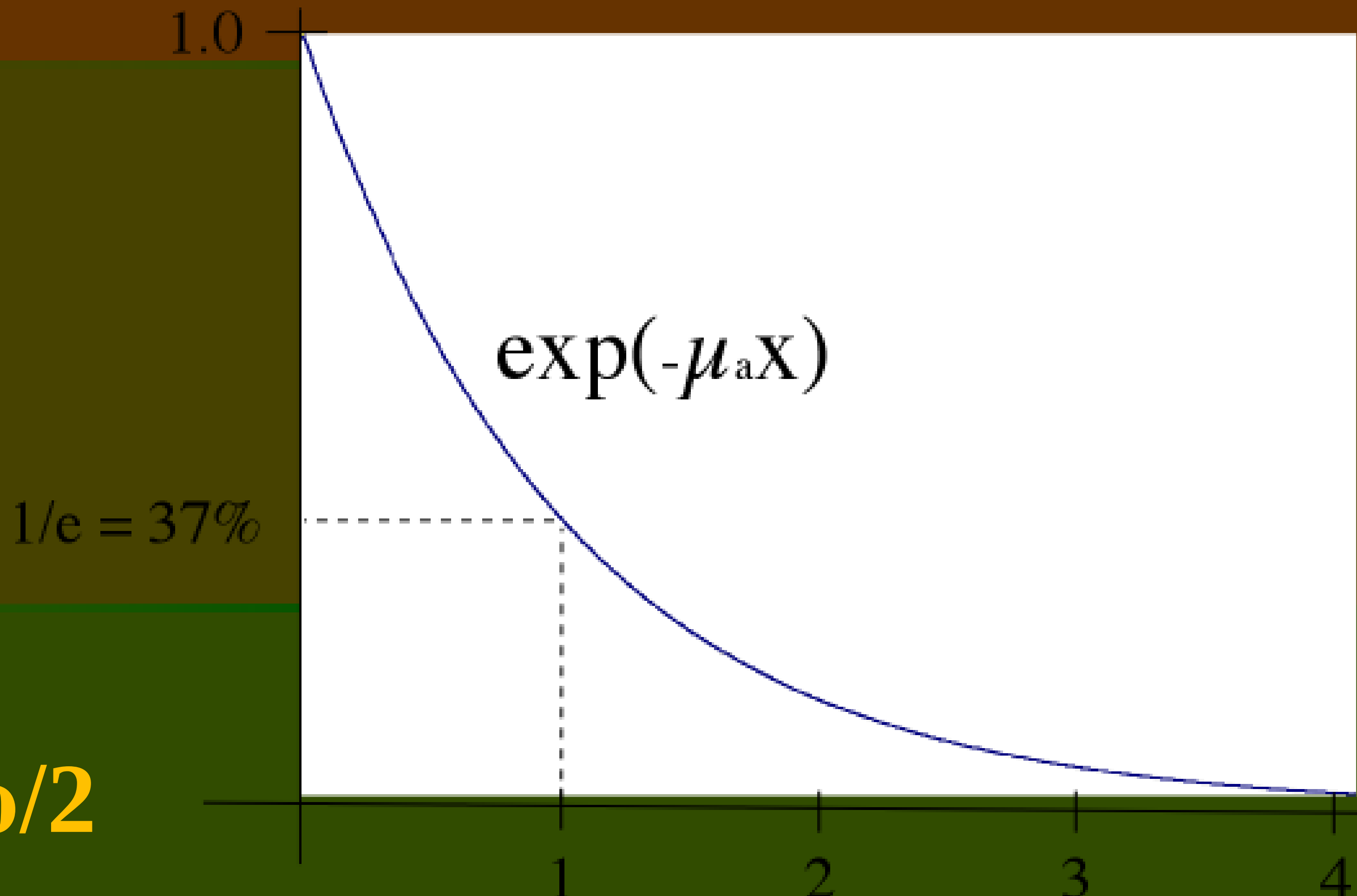


hν = 0.511 MeV
hν = 0.511 MeV
e+
hν (>1.02 MeV)
nucleus
e−

For example if a photon with energy *3Mev* is interacted with the nucleus, *1.02Mev* is used to creat the electron and the positron, the rest energy *1.98 Mev* is equally distributed between the two electrons as kinetic energy each electron has *0.99 Mev* . These two particles lose their energy by ionization, until a positron (antiparticle of the electron) is annihilated by an electron with the production of two photons.

Attenuation Law

$$I / I_o = e^{-\mu X}$$

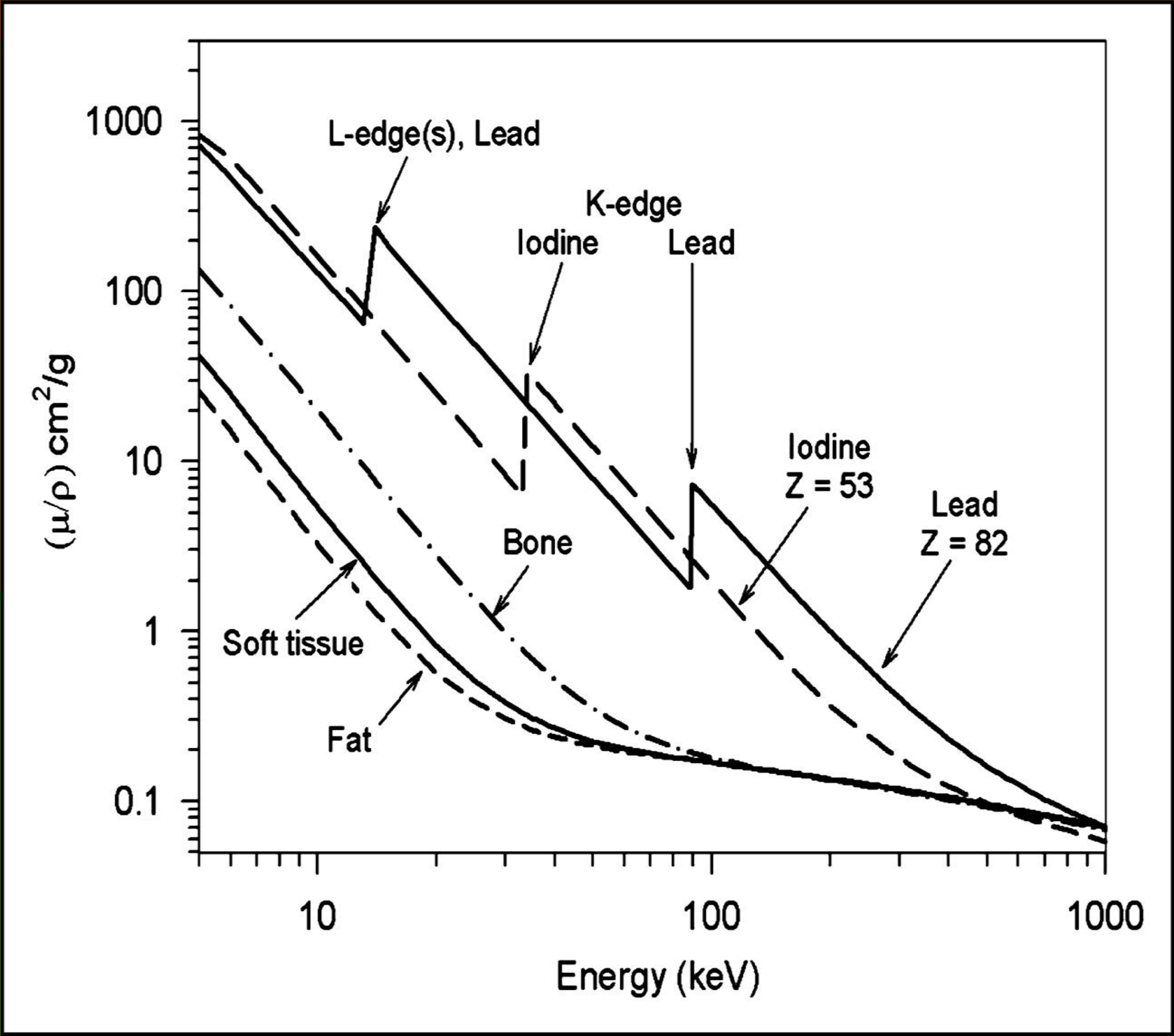


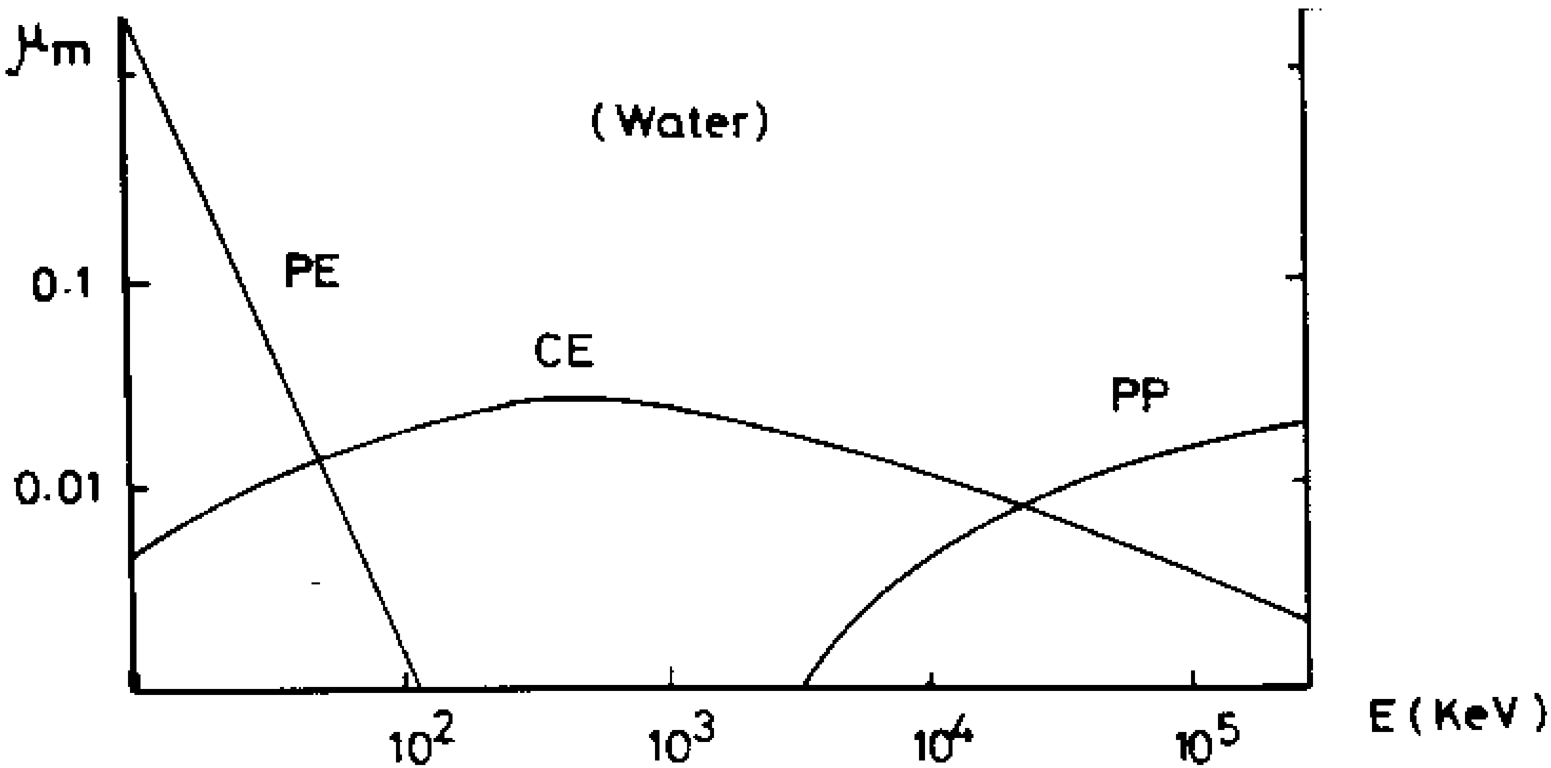
at $I = I_o/2$

$X = X_{½}$ $\quad X_{½} = 0.693/\mu$

Mass attenuation coefficient for various tissues , lead , iodine

sharp rises are called K or L– edges.





μm
(Water)
PE
CE
PP
0.1
0.01
10^2
10^3
10^4
10^5
E (KeV)

$$H_2O \xrightarrow{Rad} H_2O^+ + e^-$$

$$H_2O^+ \rightarrow H^+ + OH$$

$$H_2O + e^- \rightarrow H_2O^-$$

$$H_2O^- \rightarrow H + OH^-$$

$$OH + OH \rightarrow H_2O_2$$

$$H + O_2 \rightarrow HO_2$$

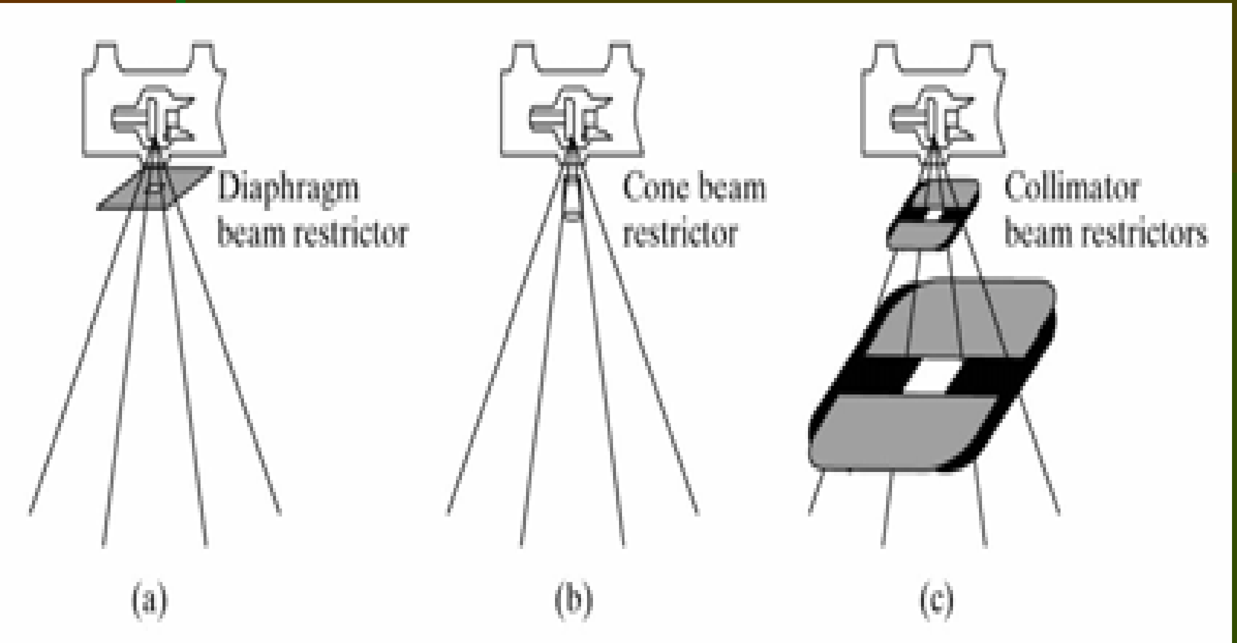
γ OR X – RAY IMAGING

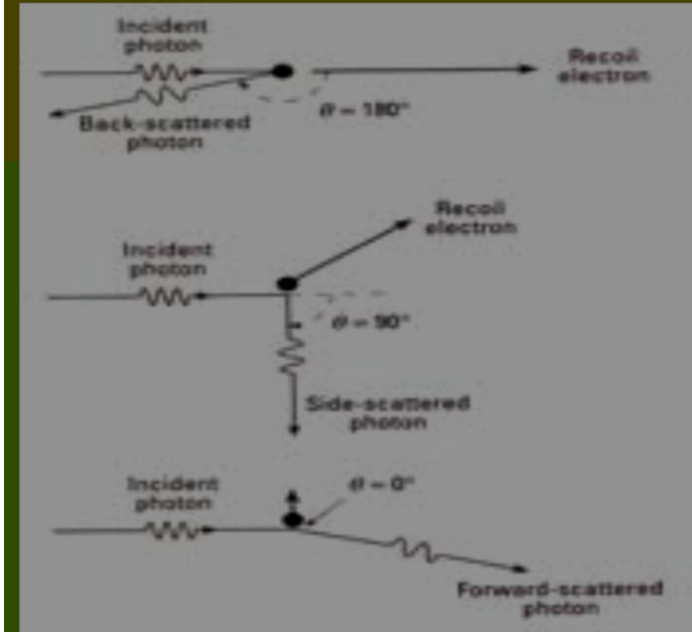
Body tissues attenuate photons in different amounts depending on their linear attenuation coefficients and photons energies.

This differential attenuation gives rise to contrast, and therefore the ability to differentiate tissues.

Since γ or X – ray is invisible; this image is converted to visible one by fluorescent screen, image intensifier, or film. The beam that emerges from the patient contains primary and scattered radiation. Since the primary beam contains useful information about the object being examined ,it is desirable to reduce the amount radiation to the film by means of grid.

This grid allows only that radiation which comes from the direction of the source to reach the imaging system. The scatter component may be also reduced by limiting the size of the X – ray field to the region to be examined.





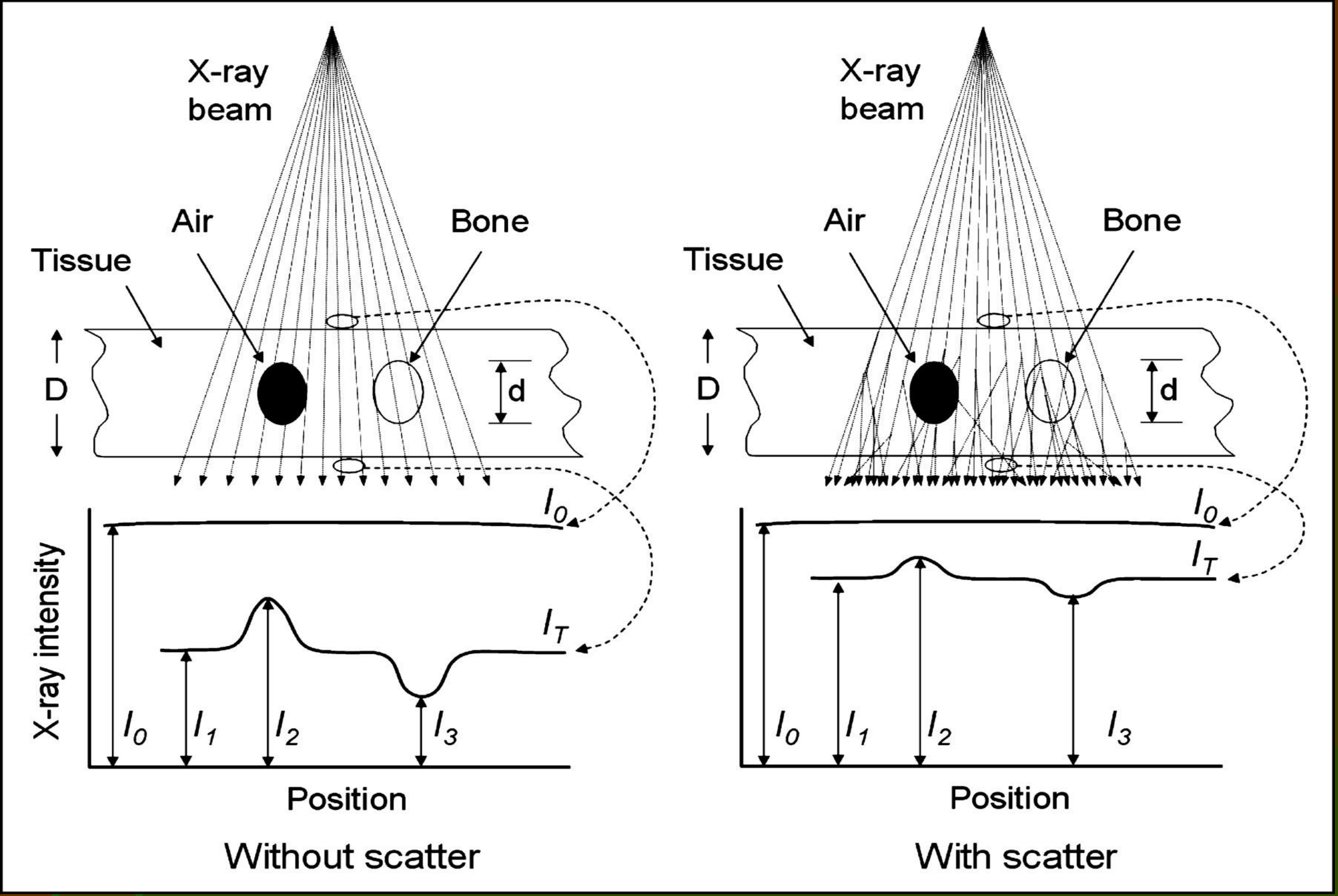
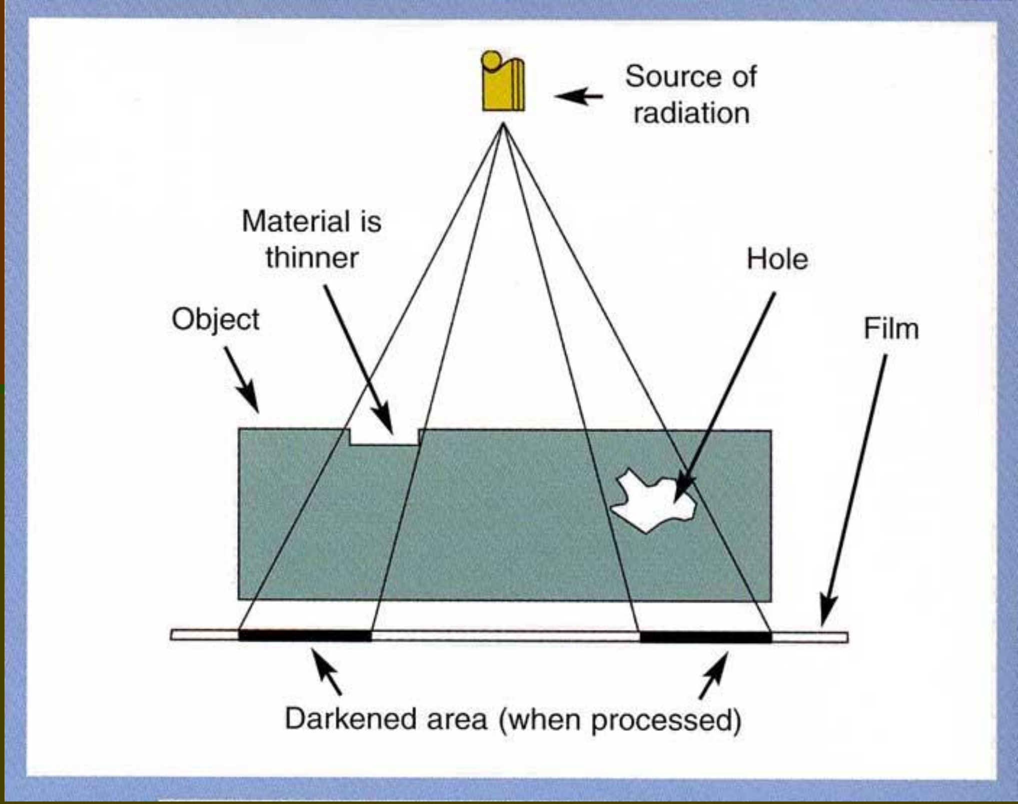


Image production with X – rays .



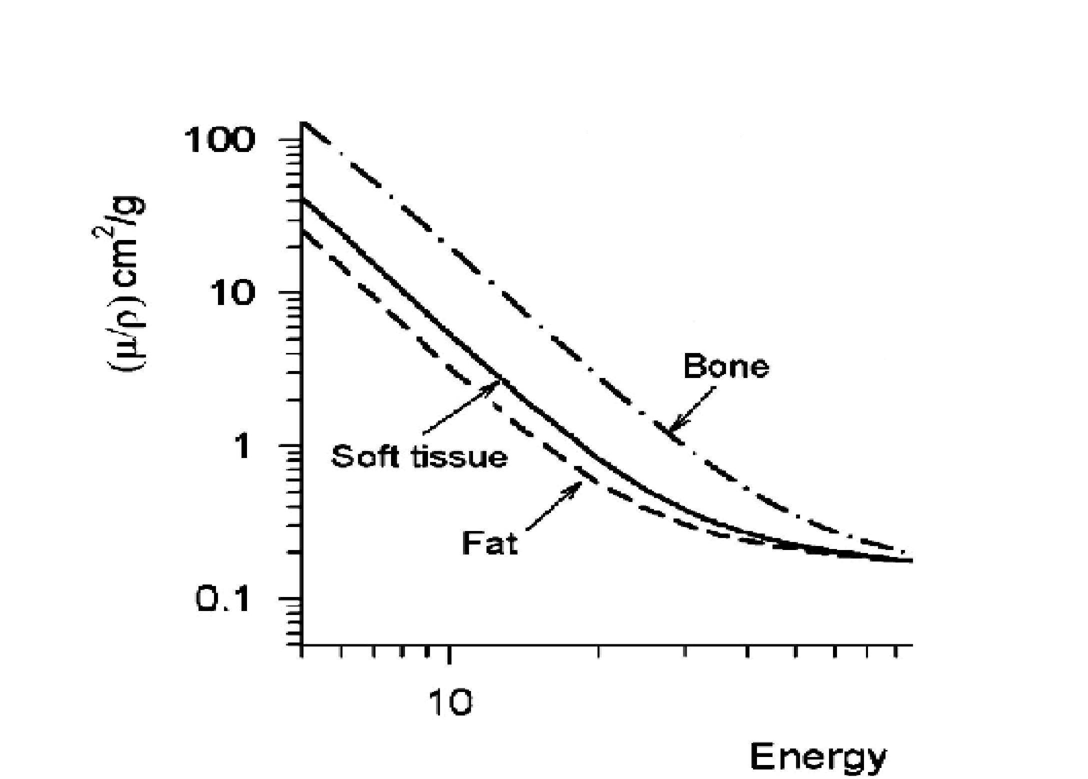
Imaging using γ source

Images Produced by Tissues

Fat, muscles and bone can be distinguished from one another in an X – rays in different ways.

Contrast in x-ray imaging.
Large contrast between bone and muscle Decreases with increasing energy (photoelectric absorption in bone is much larger at low energy)

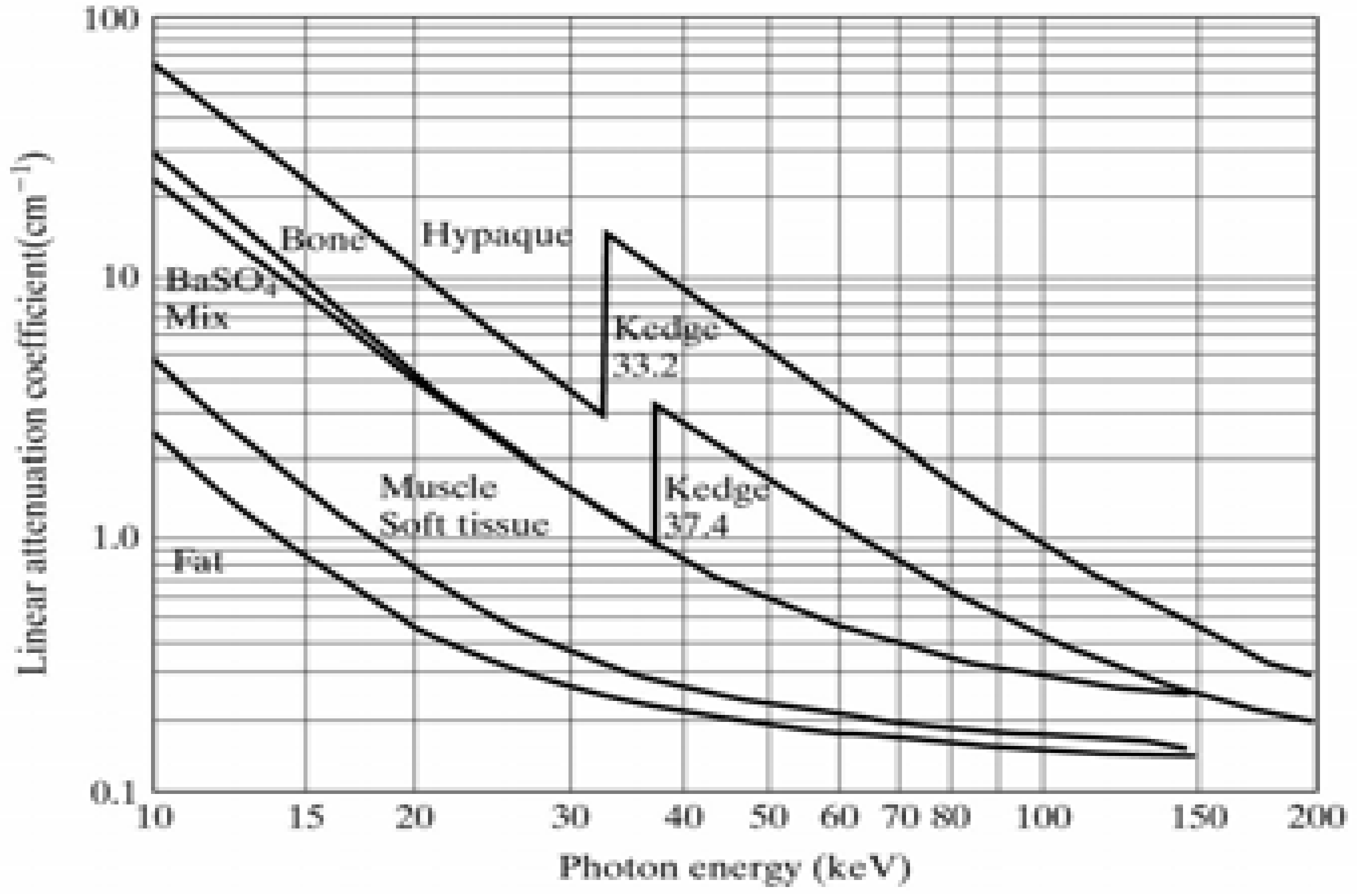
- **Little contrast between muscle and other soft tissue**
- **Large difference between air and tissue (owing to difference in density)**



Contrast agents

The two most common contrast agents are iodine (Z=53) and barium (Z=56).

These agents have relatively high atomic numbers, and they have K-shell electrons whose binding energies fall within the diagnostic x-ray energy range (30 - 40 keV)

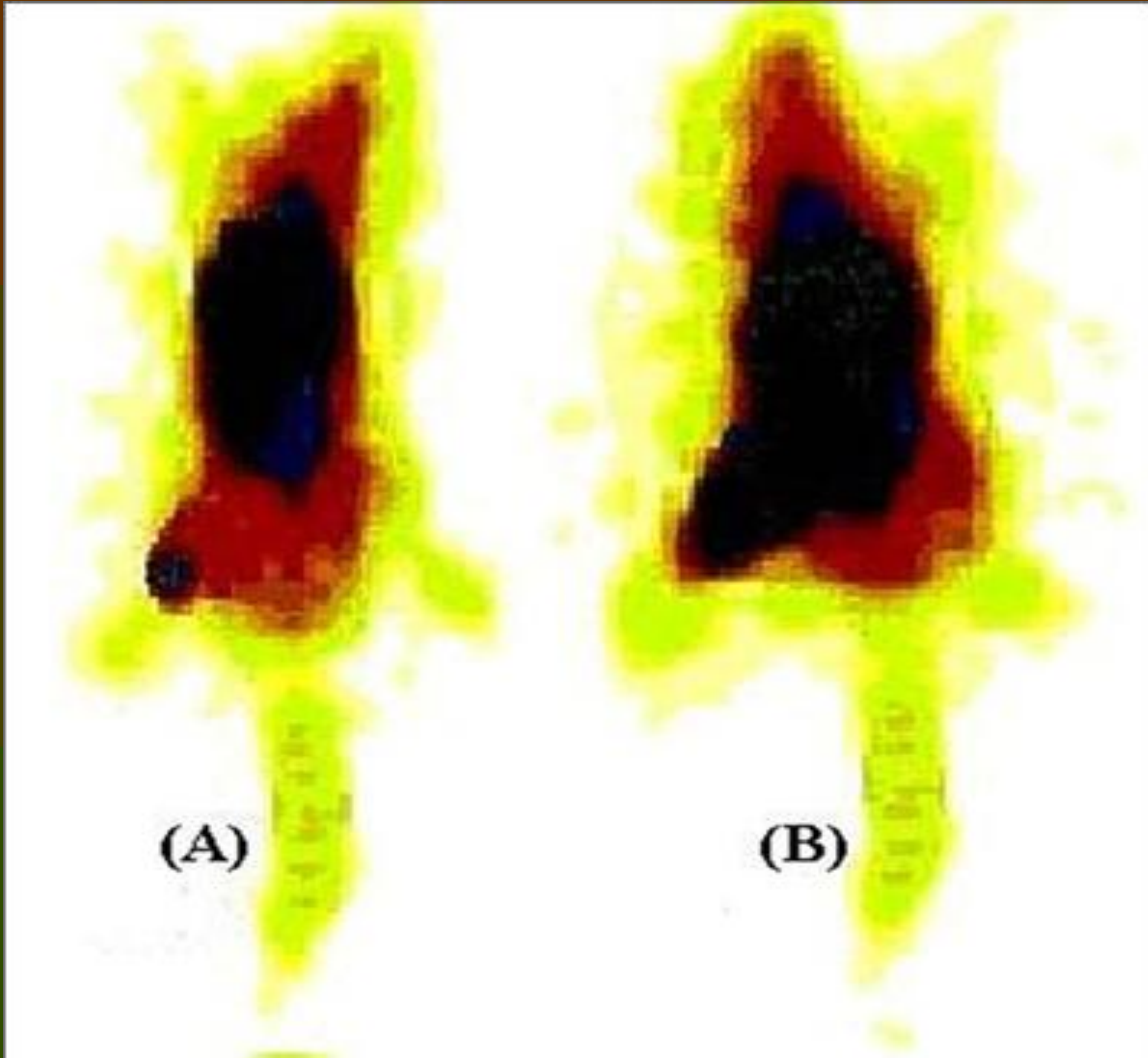


Iodine will cast a very dense shadow especially for energies just above its K – edge at 33.2 Kev, while barium shows the same effect just above 37.4 Kev.We

Images Produced by Contrast Media

To visualize many organs in the body it is necessary to introduce into the patient a medium that is deposited in the organ and that absorbs X – rays either more or less than surrounding tissues.

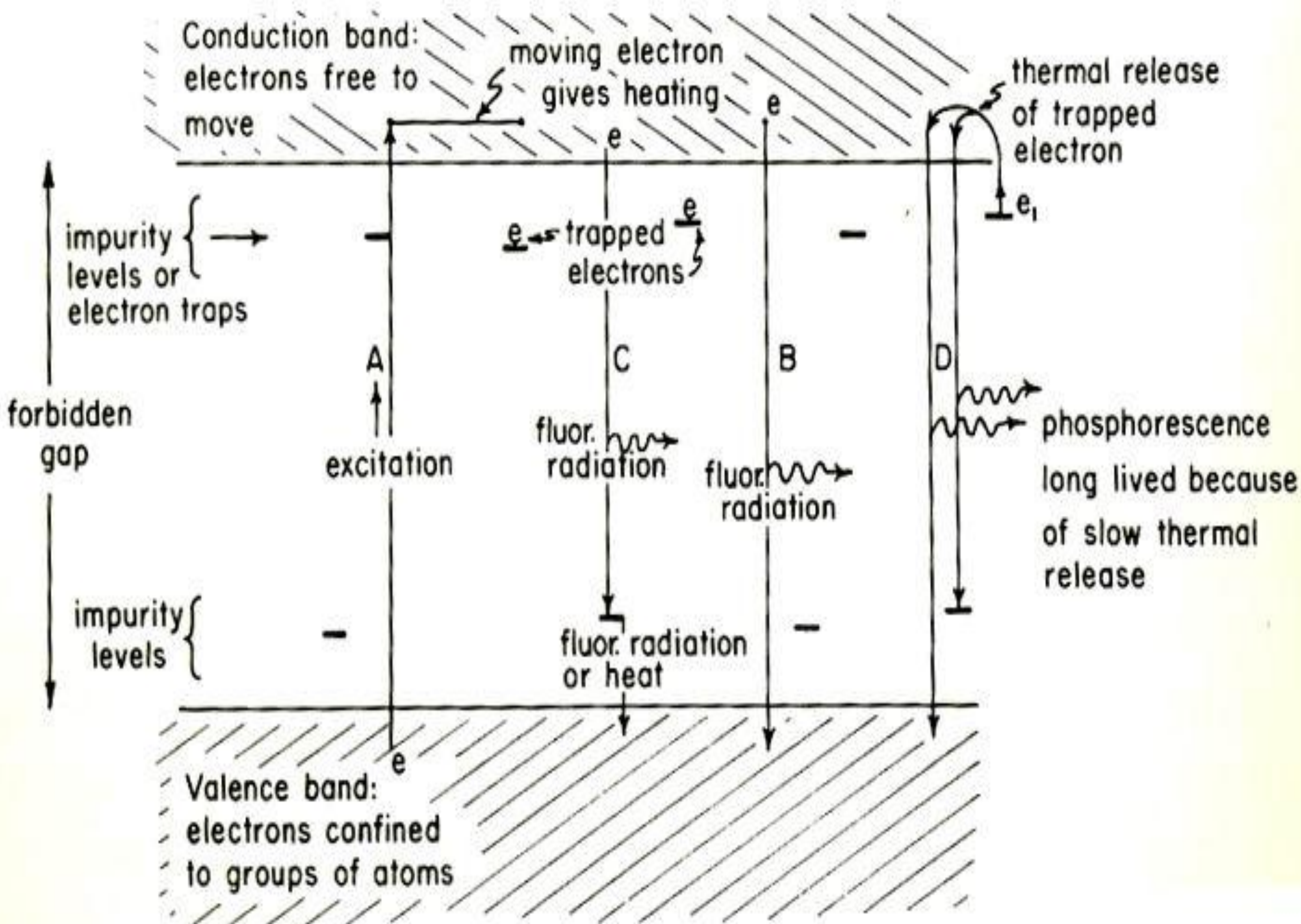
Barium and Iodine absorb X- rays very strongly , and so organs filled with these agents transmit very little radiation . The absorption data for barium and iodine are plotted in figure for suspensions of these materials as they are commonly used. Both show a sharp discontinuity in the adsorption curve at the K – edge.



(A)
(B)

PHOSPHORS
(FLUORESCENT MATERIALS)

The conversion of X – ray energy to light makes use of the fluorescence and phosphorescence of certain crystals by the mechanism illustrated in figure .



Conduction band:
electrons free to
move
moving electron
gives heating
e
e
thermal release
of trapped
electron
e_1
impurity
levels or
electron traps
e
trapped
e
electrons
A
C
B
D
forbidden
gap
excitation
fluor.
radiation
fluor.
radiation
phosphorescence
long lived because
of slow thermal
release
impurity
levels
fluor. radiation
or heat
e
Valence band:
electrons confined
to groups of atoms

the possible energy levels for the electrons are a continuous band or what is called valence band . If these electrons are excited by X–rays, they can move up to the conduction band where the electrons will be free.

Electrons cannot exist in the forbidden gap. However, if a pure crystal contains small amount of impurity discrete positions for electrons are produced in the forbidden gap. Impurity levels may trap electrons or release them to the conduction band by thermal agitation

RADIOGRAPHIC IMAGES

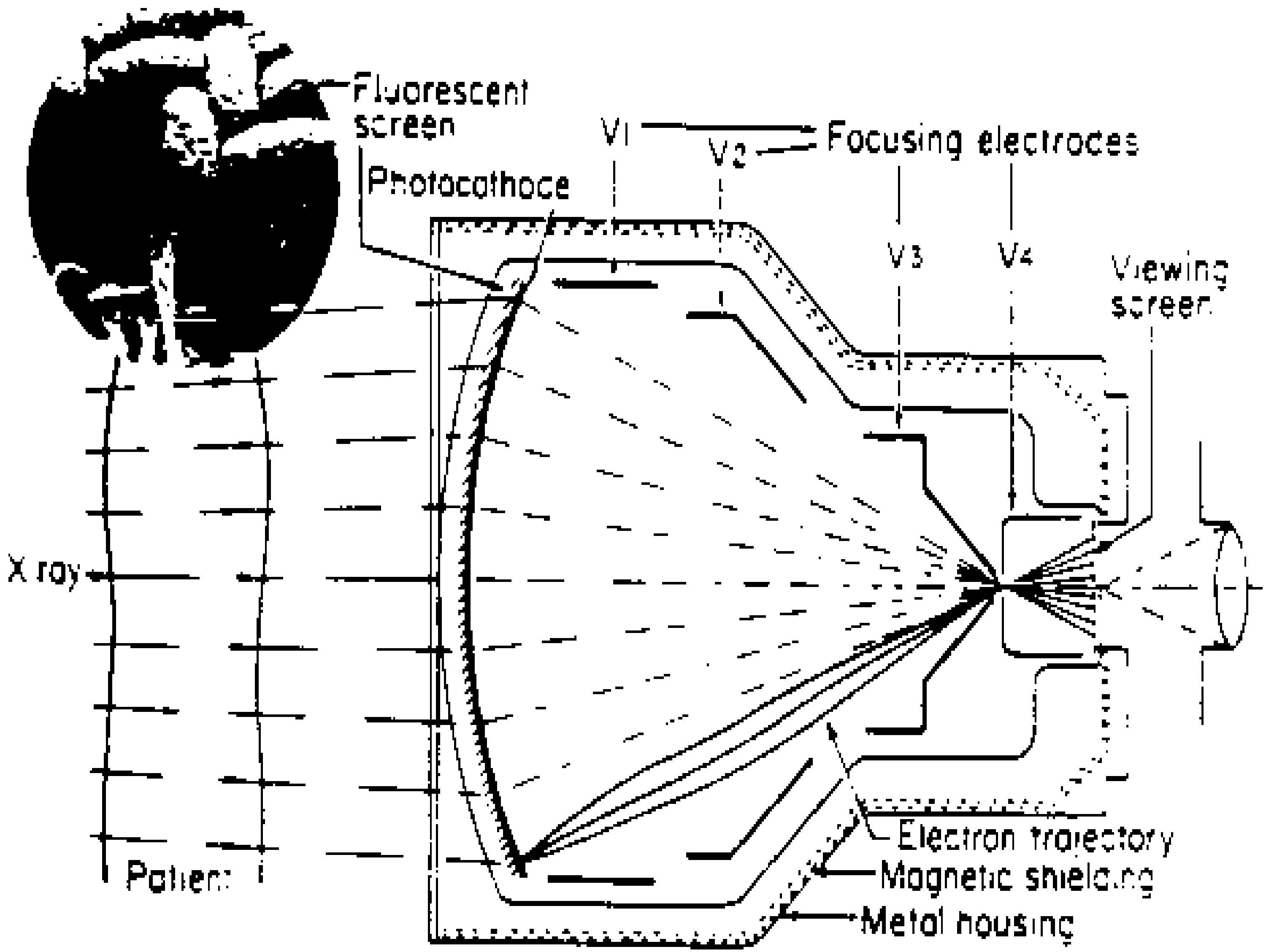
The primary radiological image described in the last section includes the information concerning the patient the radiologist would like to obtain.

However , this information is not in a useful form , and it is necessary to convert it by some type of imaging device to make it visible to radiologist this can be done by producing a visible image on the fluorescent screen of an image intensifier.

Tube Image Intensifier

An intensifier is illustrated schematically in figure. X – rays from the source passes through the patient and enters the evacuated intensifier tube through the glass or metal envelope and aluminum screen on which is deposited fluorescent material.

This material absorbs some 60% to 70% of the X – rays, converting their energy into fluorescent light. This light is absorbed by photocathode, which ejects low energy photoelectron into the evacuated tube. In a typical absorption of one X – ray photon about 100 photoelectrons are emitted.


Fluorescent screen
Photocathoce
V1
V2
Focusing electroces
V3
V4
Viewing screen
X ray
Patient
Electron trajectory
Magnetic shielding
Metal housing

These photoelectrons are accelerated and focused by the high voltage between the two ends of the tube to form a very bright image on the output phosphor or viewing screen. This image may be some 10,000 times brighter than the image on the input fluorescent screen for two reasons.

1- In the first place , the image may be reduced in diameter from about 25 cm to 2.5 cm , i.e. by a factor of 10 in diameter or of 100 in area .

Since the same number of electrons reach the output phosphor as leave the input phosphor, the number of electrons per unit area striking the output phosphor will be increased by 100 times .

2- In addition these electrons are accelerated to a high energy within the tube and on striking the output phosphor will each produce 100 times as many light photons, hence the total brightness gain is about 100 x 100 = 10,000.

The viewing screen may be observed by an optical system that enlarges it for viewing to about the same size as the original. In this process no loss in brightness; the optical system will transmit all light.

There some demands to yield the maximum signal for a given exposure :

1 – The fluorescent screen should be thick enough to absorb most of the X – ray and yet thin enough to insure that the light from the screen does not spread much before reaching the photocathode. This could be achieved by growing needlelike crystals CsI on aluminum substrate.

With this structure CsI can be made thick enough to absorb most of the radiation and yet the light generated is carried to the photocathode without spreading by the needlelike crystals, which acts as light pipes. With such a design one achieves both excellent absorption of X – rays and good resolution.

2 – The photocathode should be chosen to efficiently convert the light from the screen into photoelectron.

A structure in a patient is visible by: The *resolution* and *sharpness* of the image. The *contrast* between it and adjacent tissues caused by differences in the transmission of photons.

64 x 64

256 x 256

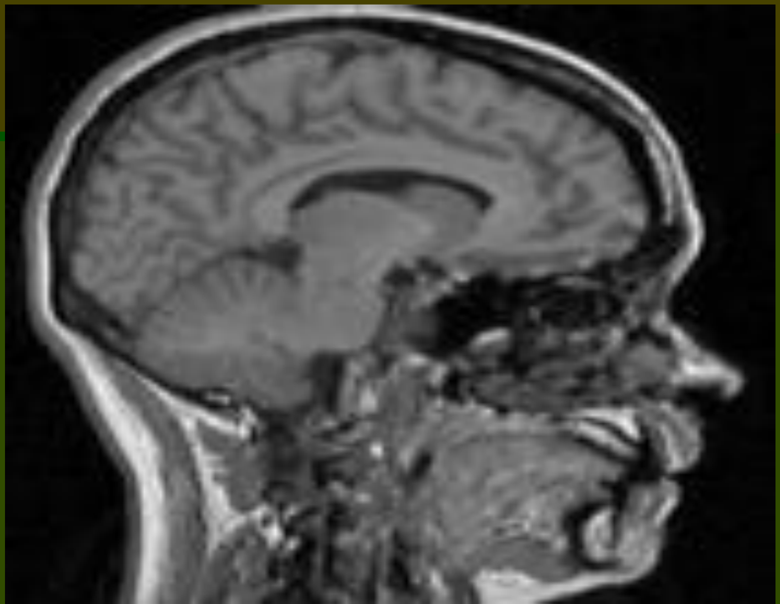
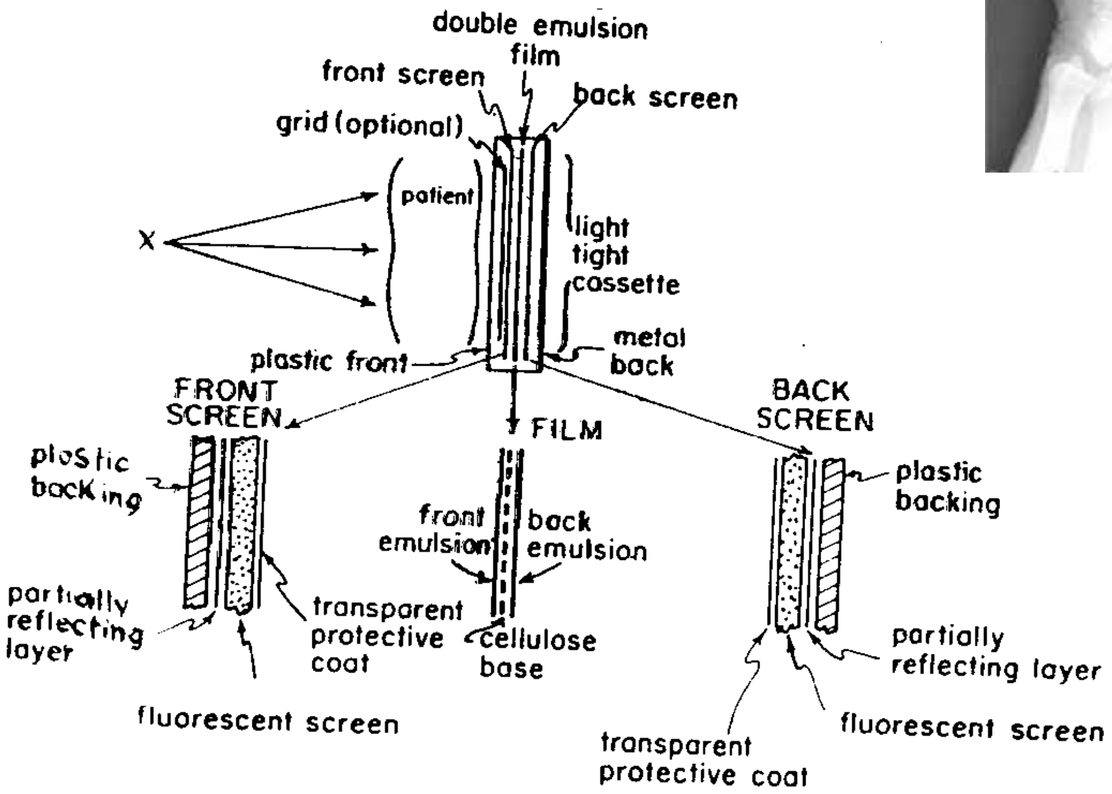


Image Produced on Full Size Film



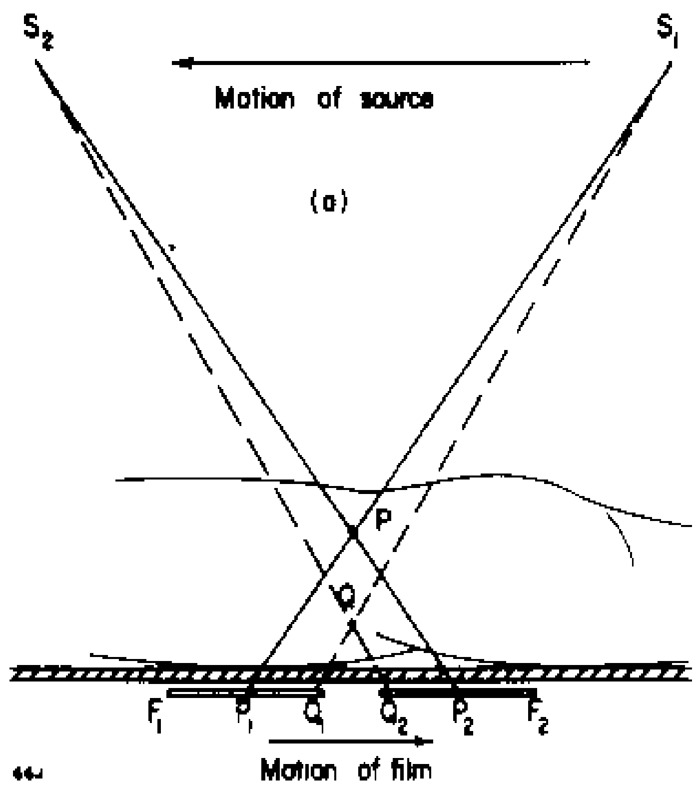
To increase the sensitivity the film is sandwiched between two fluorescent screens

This screen absorbs X-rays and emits visible and ultraviolet light, which exposes the film. The film consists of a plastic base coated on both front and back with relatively thick emulsion.

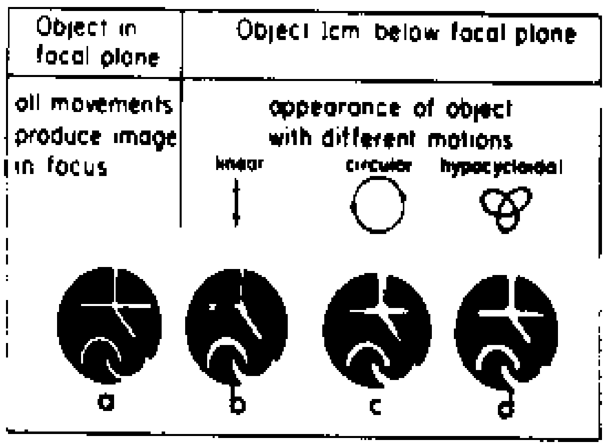
The front emulsion is exposed mainly by light from the front screen, and the back screen. Crossover: some light from the front screen reaches the back emulsion and some light from the back screen reaches the front emulsion.

TOMOGRAPHAY

A



B



Tomographic blurring movements

The X – ray source is moved in one direction from S_1 to S_2 while the film is moved in the opposite direction from P_1 to P_2 at a speed such that a line joining the center of the film to the source passes through P the center of the region of interest within the patient .

From the diagram, it is clear that the image of P will clear at P1 the center of the film for all positions of the film . On the other hand, structures in the other plane as Q will produce a blurred image extending from Q1 near one end of the film to Q2 at the other end .

The greater the distance, or the angle of movement of the X – ray tube, the narrower the slice that is in focus the greater is the blurring of overlying structures.